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HERMETIC PACKAGES AND FEEDTHROUGHS FOR NEURAL PROSTHESES

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THIS QPR IS BEING SENT TO YOU BEFORE IT HAS BEEN REVIEWED BY THE STAFF OF THE NEURAL PROSTHESIS PROGRAM.

SUMMARY

During the past quarter we continued testing of glass packages under accelerated conditions, continued the characterization of the receiver circuitry for the single-channel microstimulator and assembled a few microstimulators using the old circuit chips, and completed the layout of a fully integrated multichannel nerve stimulation system and began the fabrication of these devices.

Our most significant package testing results to date are those obtained from a series of silicon-glass packages that have been soaking in DI water at 85°C and 95°C for more than a year. We reported in previous progress reports that all packages soaking at 95°C had failed. There were also 4 packages that were soaking at 85°C. All these four packages are still dry and under test. Of the original 10 packages, the longest going sample has reached a maximum of 1104 days at 85°C and 484 days at 95°C. If we assume that all of the packages at 85°C failed the same time that the 95°C packages failed, we can calculate a worst case mean time to failure of 556 days for the samples at 85°C, and of 119 days for the samples soaking at 95°C. The worst case MTTF at body temperature based on these tests is then calculated to be 59 years. These tests have been very encouraging and clearly indicate the packages can last for many years in water. In addition to these tests in DI water, we had also soaked several packages in ***saline*** at the above two temperatures. The results obtained from these tests were reported in the last progress report. We also have had 4 packages soaking at room temperature in saline. The longest lasting package has been soaking for 1005 days, and an average soak period of 829 days at room temperature. We will continue to observe these packages for any sign of leakage. A new set of package tests were also initiated this past quarter at 85 and 95°C. At each temperature, eight packages that have been coated with silicone to prevent premature dissolution of the top polysilicon bonding layer have now been soaking for 40 days each. Only one package at 85°C failed after the first day due to premature failure and possible damage in the bond area. All the other packages have now stayed dry for the duration of the soak.

In-vivo testing of these packages continued at Michigan, Johns Hopkins University, and Vanderbilt University by Dr. Zealear's group. All of these tests indicate that the silicon-glass package is indeed biocompatible and do not induce any damage in the surrounding tissue. A detailed description of these results will be presented in the next progress report once all the data is available.

Finally, during the past quarter we completed the layout and complete simulation of the circuitry for a fully integrated 8-channel mini-microstimulator that can operate using on-chip coils. This circuitry can be used for peripheral nerve stimulation applications and can deliver current pulses to any of 8 electrodes. The circuitry was layed out and sent out for masks. Masks have been received and fabrication of these devices has already started. The mask set also contains designs for the single-channel and multi-channel microstimulator which we will use to assembly functional devices for our collaborators at Vanderbilt University and Hines VA hospital. We believe that fabrication will be complete by the end of the next quarter. We also further characterized on-chip coils for power and data reception and are confident that sufficient levels of power can be transmitted to these on-chip coils for peripheral nerve applications. The mini-microstimulator can thus operate without the need for any hybrid components, which will make the fabrication and assembly of these systems much easier.

1. INTRODUCTION

This project deals with the development of hermetic, biocompatible micropackages and feedthroughs for use in a variety of implantable neural prostheses for sensory and motor handicapped individuals. The project also aims at continuing work on the development of a telemetrically powered and controlled neuromuscular microstimulator for functional electrical stimulation. The primary objectives of the project are: 1) the development and characterization of hermetic packages for miniature, silicon-based, implantable three-dimensional structures designed to interface with the nervous system for periods of up to 40 years; 2) the development of techniques for providing multiple sealed feedthroughs for the hermetic package; 3) the development of custom-designed packages and systems used in chronic stimulation or recording in the central or peripheral nervous systems in collaboration and cooperation with groups actively involved in developing such systems; and 4) establishing the functionality and biocompatibility of these custom-designed packages in *in-vivo* applications. Although the project is focused on the development of the packages and feedthroughs, it also aims at the development of inductively powered systems that can be used in many implantable recording/stimulation devices in general, and of multichannel microstimulators for functional neuromuscular stimulation in particular.

Our group here at the Center for Integrated Sensors and Circuits at the University of Michigan has been involved in the development of silicon-based multichannel recording and stimulating microprobes for use in the central and peripheral nervous systems. More specifically, during the past two contract periods dealing with the development of a single-channel inductively powered microstimulator, our research and development program has made considerable progress in a number of areas related to the above goals. A hermetic packaging technique based on electrostatic bonding of a custom-made glass capsule and a supporting silicon substrate has been developed and has been shown to be hermetic for a period of at least a few years in salt water environments. This technique allows the transfer of multiple interconnect leads between electronic circuitry and hybrid components located in the sealed interior of the capsule and electrodes located outside of the capsule. The glass capsule can be fabricated using a variety of materials and can be made to have arbitrary dimensions as small as 1.8mm in diameter. A multiple sealed feedthrough technology has been developed that allows the transfer of electrical signals through polysilicon conductor lines located on a silicon support substrate. Many feedthroughs can be fabricated in a small area. The packaging and feedthrough techniques utilize biocompatible materials and can be integrated with a variety of micromachined silicon structures.

The general requirements of the hermetic packages and feedthroughs to be developed under this project are summarized in Table 1. Under this project we will concentrate our efforts to satisfy these requirements and to achieve the goals outlined above. There are a variety of neural prostheses used in different applications, each having different requirements for the package, the feedthroughs, and the particular system application. The overall goal of the program is to develop a miniature hermetic package that can seal a variety of electronic components such as capacitors and coils, and integrated circuits and sensors (in particular electrodes) used in neural prostheses. Although the applications are different, it is possible to identify a number of common requirements in all of these applications in addition to those requirements listed in Table 1. The packaging and feedthrough technology should be capable of:

1- protecting non-planar electronic components such as capacitors and coils, which typically have large dimensions of about a few millimeters, without damaging them;
2- protecting circuit chips that are either integrated monolithically or attached in a hybrid fashion with the substrate that supports the sensors used in the implant;
3- interfacing with structures that contain either thin-film silicon microelectrodes or conventional microelectrodes that are attached to the structure;

Table 1: General Requirements for Miniature Hermetic Packages and Feedthroughs for Neural Prostheses Applications

Package Lifetime:

≥ 40 Years in Biological Environments @ 37°C

Packaging Temperature:

≤360°C

Package Volume:

10-100 cubic millimeters

Package Material:

Biocompatible
Transparent to Light
Transparent to RF Signals

Package Technology:

Batch Manufactureable

Package Testability:

Capable of Remote Monitoring
In-Situ Sensors (Humidity & Others)

Feedthroughs:

At Least 12 with ≤125μm Pitch
Compatible with Integrated or Hybrid Microelectrodes
Sealed Against Leakage

Testing Protocols:

In-Vitro Under Accelerated Conditions
In-Vivo in Chronic Recording/Stimulation Applications

We have identified two general categories of packages that need to be developed for implantable neural prostheses. The first deals with those systems that contain large components like capacitors, coils, and perhaps hybrid integrated circuit chips. The second deals with those systems that contain only integrated circuit chips that are either integrated in the substrate or are attached in a hybrid fashion to the system.

Figure 1 shows our general proposed approach for the package required in the first category. This figure shows top and cross-sectional views of our proposed approach here. The package is a glass capsule that is electrostatically sealed to a support silicon substrate. Inside the glass capsule are housed all of the necessary components for the system. The electronic circuitry needed for any analog or digital circuit functions is either fabricated on a separate circuit chip that is hybrid mounted on the silicon substrate and electrically connected to the silicon substrate, or integrated monolithically in the support silicon substrate itself. The attachment of the hybrid IC chip to the silicon substrate can be performed using a number of different technologies such as simple wire bonding between pads located on each substrate, or using more sophisticated techniques such as flip-chip solder reflow or tab bonding. The larger capacitor or microcoil components are mounted on either the substrate or the IC chip using appropriate epoxies or solders. This completes the assembly of the electronic components of the system and it should be possible to test the system electronically at this point before the package is completed. After testing, the system is packaged by placing the glass capsule over the entire system and bonding it to the silicon substrate using an electrostatic sealing process. The cavity inside the glass package is now hermetically sealed against the outside environment. Feedthroughs to the outside world are provided using the grid-feedthrough technique discussed in previous reports. These feedthroughs transfer the electrical signals between the electronics inside the package and various elements outside of the package. If the package has to interface with conventional microelectrodes, these microelectrodes can be attached to bonding pads located outside of the package; the bond junctions will have to be protected from the external environment using various polymeric encapsulants. If the package has to interface with on-chip electrodes, it can do so by integrating the electrode on the silicon support substrate. Interconnection is simply achieved using on-chip polysilicon conductors that make the feedthroughs themselves. If the package has to interface with remotely located recording or stimulating electrodes that are attached to the package using a silicon ribbon cable, it can do so by integrating the cable and the electrodes again with the silicon support substrate that houses the package and the electronic components within it.

Figure 2 shows our proposed approach to package development for the second category of applications. In these applications, there are no large components such as capacitors and coils. The only component that needs to be hermetically protected is the electronic circuitry. This circuitry is either monolithically fabricated in the silicon substrate that supports the electrodes (similar to the active multichannel probes being developed by the Michigan group), or is hybrid attached to the silicon substrate that supports the electrodes (like the passive probes being developed by the Michigan group). In both of these cases the package is again another glass capsule that is electrostatically sealed to the silicon substrate. Notice that in this case, the glass package need not be a high profile capsule, but rather it need only have a cavity that is deep enough to allow for the silicon chip to reside within it. Note that although the silicon IC chip is originally 500μm thick, it can be thinned down to about 100μm, or can be recessed in a cavity created in the silicon substrate itself. In either case, the recess in the glass is less than 100μm deep (as opposed to several millimeters for the glass capsule). Such a glass package can be easily fabricated in a batch process from a larger glass wafer.

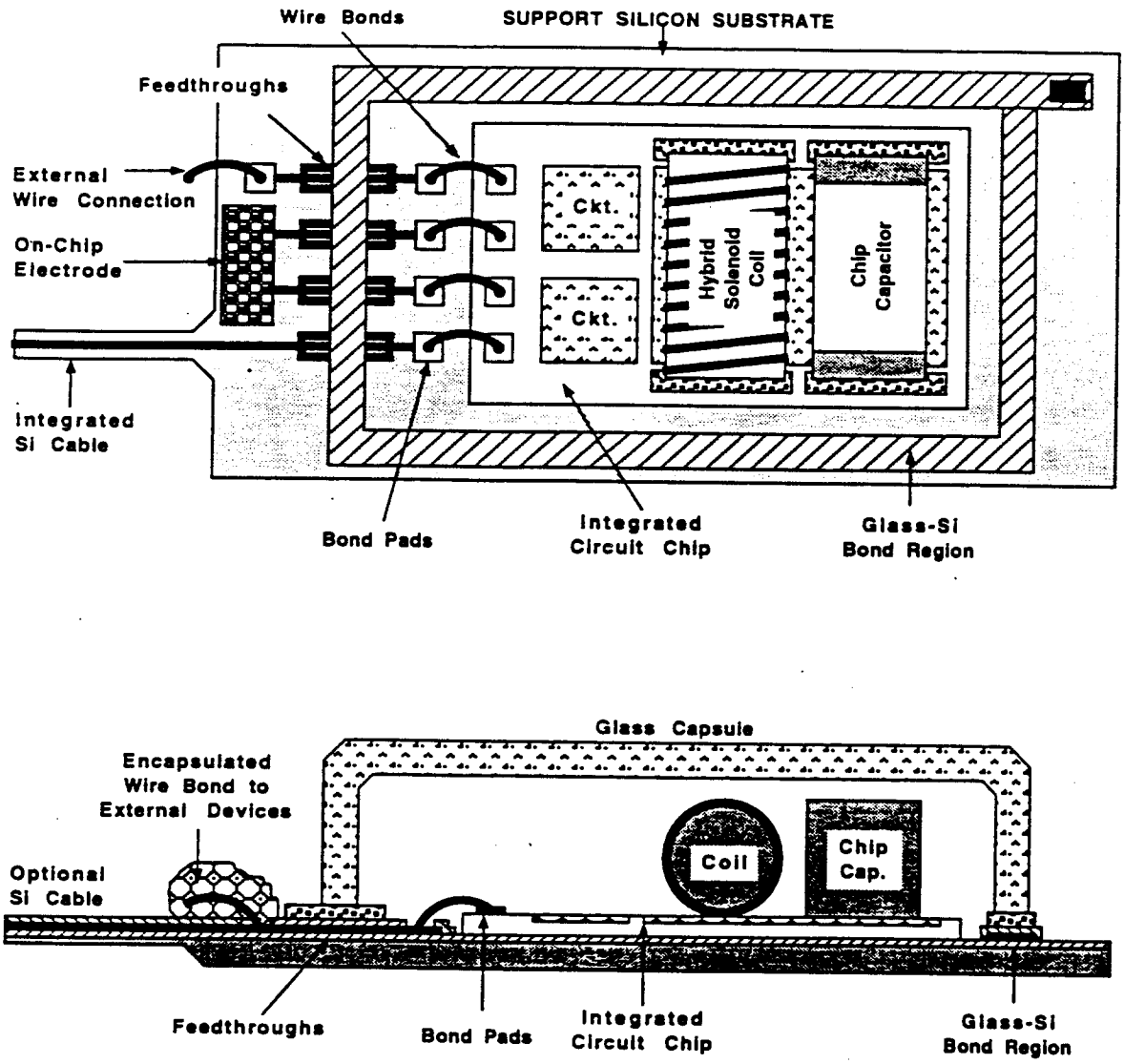

Figure 1: A generic approach for packaging implantable neural prostheses that contain a variety of components such as chip capacitors, microcoils, and integrated circuit chips. This packaging approach allows for connecting to a variety of electrodes.

We believe the above two approaches address the needs for most implantable neural prostheses. Note that both of these techniques utilize a silicon substrate as the supporting base, and are not directly applicable to structures that use other materials such as ceramics or metals. Although this may seem a limitation at first, we believe that the use of silicon is, in fact, an advantage because it provides several benefits. First, it is biocompatible and has been used extensively in biological applications. Second, there is a great deal of effort in the IC industry in the development of multi-chip modules (MCMs), and many of these efforts use silicon supports because of the ability to form high density interconnections on silicon using standard IC fabrication techniques. Third, many present and future implantable probes are based on silicon micromachining technology; the use of our proposed packaging technology is inherently compatible with most of these probes, which simplifies the overall structure and reduces its size.

Once the above packages are developed, we will test them in biological environments by designing packages for specific applications. One of these applications is in recording neural activity from cortex using silicon microprobes developed by the Michigan group under separate contracts. The other involves the chronic stimulation of muscular tissue using a multichannel microstimulator for the stimulation of the paralyzed larynx. This application has been developed at Vanderbilt University. Once the device is built, it will be used by our colleagues at Vanderbilt to perform both biocompatibility tests and functional tests to determine package integrity and suitability and device functionality for the reanimation of the paralyzed larynx. The details of this application will be discussed in future progress reports.

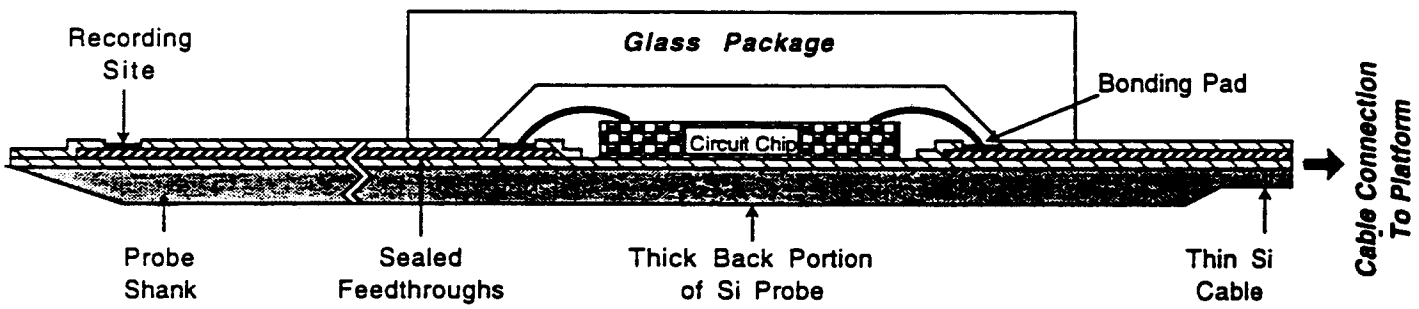

Figure 2: Proposed packaging approach for implantable neural prostheses that contain electronic circuitry, either monolithically fabricated in the probe substrate or hybrid attached to the silicon substrate containing microelectrodes.

2. ACTIVITIES DURING THE PAST QUARTER

2.1 Hermetic Packaging

Over the past few years we have developed a bio-compatible hermetic package with high density multiple feedthroughs. This package utilizes electrostatic bonding of a custom-made glass capsule to a silicon substrate to form a hermetically sealed cavity, as shown in Figure 3. Feedthrough lines are obtained by forming closely spaced polysilicon lines and planarizing them with LTO and PSG. The PSG is reflowed at 1100°C for 2 hours to form a planarized surface. A passivation layer of oxide/nitride/oxide is then deposited on top to prevent direct exposure of PSG to moisture. A layer of fine-grain polysilicon (surface roughness ≈50Å rms) is deposited and doped to act as the bonding surface. Finally, a glass capsule is bonded to this top polysilicon layer by applying a voltage of 2000V between the two for 10 minutes at 320 to 340°C, a temperature compatible with most hybrid components. The glass capsule can be either custom molded from Corning code #7740 glass, or can be batch fabricated using ultrasonic micromachining of #7740 glass wafers.

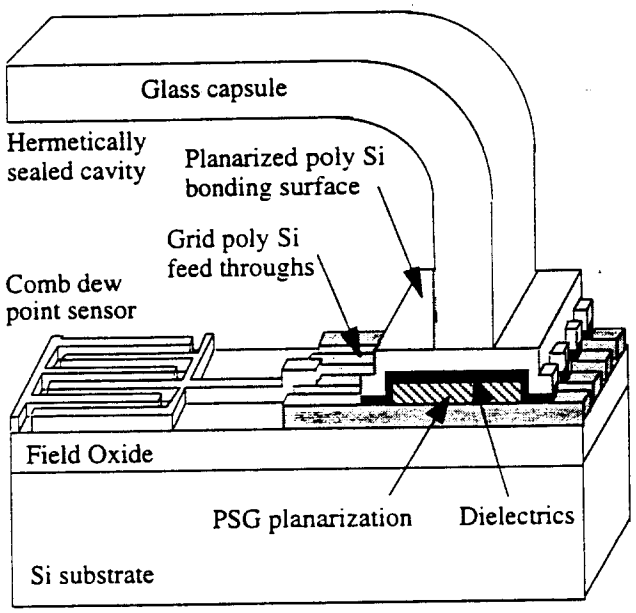

Figure 3: The structure of the hermetic package with grid feedthroughs.

During the past years we have electrostatically bonded and soak tested over one hundred and sixty of these packages. The packages successfully prevent leakage in soak tests at 95°C for over 4 months on average and at 85°C for over 18.5 months in deionized water. The bonding yield has varied between 85% to 72% (yield is defined as the percentage of packages which last more than 24 hours in the solution they are soaked in). It is worth noting that earlier tests that have been ongoing for more than about 3 years (room temperature soak tests in saline and the 85°C and the 95°C tests in deionized water) have been made with silicon substrates that are thinned (~150μm) and bonded to the custom molded glass capsules. All of the relatively recent tests (85°C and 95°C tests in saline) are performed with silicon substrates having full thickness (~500μm) and bonded to ultrasonically machined glass capsules with a flat top surface. We have also fabricated a low profile glass package with a smaller cavity that could specifically be used for the encapsulation of integrated components on a circuit chip. We have performed soak tests both with deionized water and phosphate buffered saline during the past year. Due to the enhanced dissolution rate of polysilicon in saline, tests on the samples in saline have been

concluded. We currently have devices being tested in deionized water for over 3 years at 85°C. The in-vivo test results from both the normal package and the low profile package reveal that the devices are biocompatible and rugged. We have implanted about 11 passive devices during the past quarter with 2 of them already being explanted after 1 month and the other results to be reported as they become available. We have used a wafer with evaporated glass deposited on top and a silicon substrate and have made our first all silicon package. We have also started new soak tests in saline with silicone coated samples during this past quarter which will be discussed in detail in the following section.

2.1.1 Ongoing Accelerated Soak Tests in Deionized Water

Accelerated soak tests of the glass-silicon package in DI water have continued and 20% of the packages have now surpassed three years of accelerated testing with no sign of moisture penetration. For these tests temperature is chosen as the accelerating factor since it is an easy variable to control and also the diffusion of moisture is a strong (exponential) function of temperature. We started soaking 10 samples each at 85°C and 95°C in this group of tests. Tables 2 and 3 below list some pertinent data from these soak tests. Figure 4 summarizes the final results from the 95°C soak tests and Figure 5 summarizes the results so far from the 85°C tests. In these figures we point to the causes of failure for individual packages when it is known, and also show a curve fit to our lifetime data to illustrate the general trend. Moreover, the curve fit only approximates the actual package lifetimes since some of our packages failed due to handling during testing rather than due to leakage.

Table 2: Key data for 95°C soak tests in DI water.

Number of packages in this study	10
Soaking solution	DI water
Failed within 24 hours (not included in MTTF)	1
Packages lost due to mishandling	2
Longest lasting packages in this study	484 days
Packages still under tests with no measurable room temperature condensation inside	0
Average lifetime to date (MTTF) including losses attributed to mishandling	*118.7 days*
Average lifetime to date (MTTF) not including losses attributed to mishandling	*135.7 days*

Table 3: Key data for 85°C soak tests in DI water.

Number of packages in this study	10
Soaking solution	DI water
Failed within 24 hours (not included in MTTF)	2
Packages lost due to mishandling	3
Longest lasting packages so far in this study	1104 days
Packages still under tests with no measurable room temperature condensation inside	4
Average lifetime to date (MTTF) including losses attributed to mishandling	*556 days*
Average lifetime to date (MTTF) not including losses attributed to mishandling	*873 days*

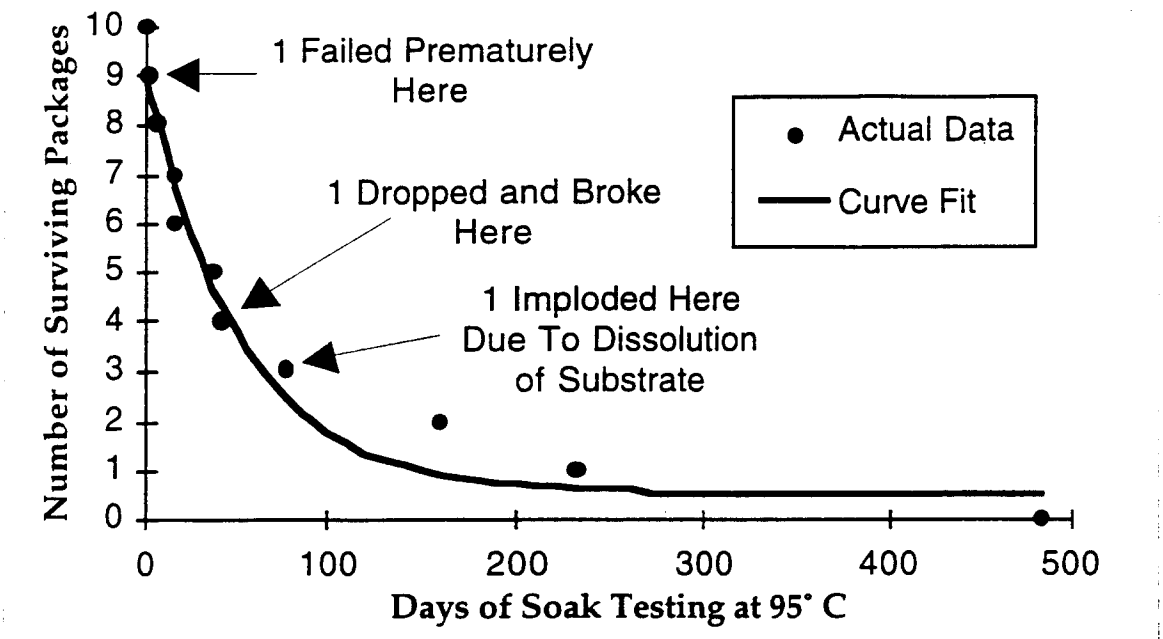

Figure 4: Summary of the lifetimes of the 10 packages which have been soak tested at 95°C in DI water.

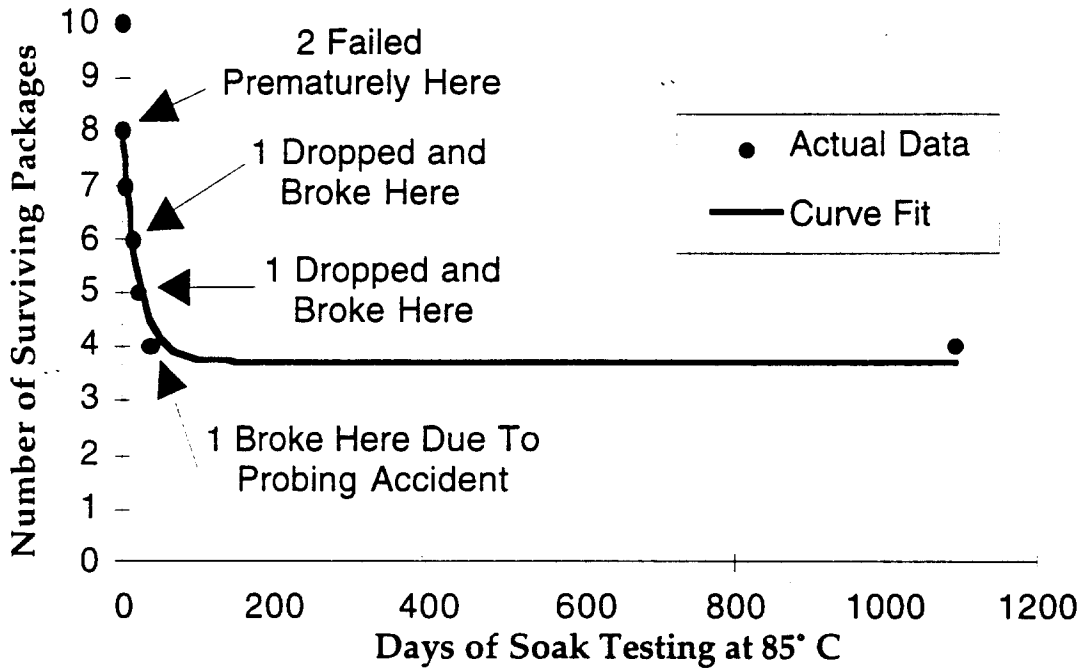

Figure 5: Summary of the lifetimes of the 10 packages which have been soak tested at 85°C in DI water.

We had 4 packages soaking at 85°C at the beginning of this quarter. All of these 4 packages have survived to the end of this quarter and are still under test. For these packages we define failure as the room temperature condensation of moisture inside the package. The testing sequence for these packages consist of cooling the sample to room temperature from their soak bath at the elevated temperature. This is followed by a DI water rinse and nitrogen dry. We next measure the impedance of the dew point sensors and inspect the sample carefully for leakage under the microscope. The significant change in impedance (about 2 orders of magnitude) and observation of visible condensation inside the package would both be classified as the failure of the package under test. Of the original 10 samples in the 95°C tests, the longest lasting package survived for a total of 484 days. The calculated mean time to failure of these packages are 135.7 days excluding the handling errors. Of the original 10 packages in the 85°C soak tests there are still four with no sign of room temperature condensation. The longest lasting package in the 85°C tests has lasted a total of 1104 days and is still under test. The worst case mean time to failure for these tests has been calculated as 873 days excluding handling errors. We have started a new set of soak tests as mentioned later in this report.

2.1.2 Interpretation of the Long Term Soak Testing Results in Deionized Water

Generally during accelerated testing, one models the mean time to failure (MTTF) as an Arrhenius processes (In the VLSI industry this model is used for failure due to diffusion, corrosion, mechanical stress, electromigration, contact failure, dielectric breakdown, and mobile ion/surface inversion). The generalized equation used in all these cases is given below where MTTF is the mean time to failure, A is a constant, ξ is the stress factor other than temperature, (such as pressure or relative humidity), n is the stress dependence, Q is the activation energy, K_B is Boltzman's constant, and T is the temperature in Kelvin.

$$MTTF = A \cdot \xi^{-n} \cdot e^{\left(\frac{Q}{K_B T}\right)}$$

For the accelerated soak tests that we have performed on the packages, there was no stressing factor other than temperature, so the ξ term drops out of the above equation. The resulting equation can be rewritten as a ratio of MTTFs as it is below. This is the model we are using to interpret the accelerated soak tests performed during the past year.

$$AF = \frac{MTTF_{Normal}}{MTTF_{Accelerated}} = e^{\frac{Q}{K_B}\left(\frac{1}{T_{Normal}} - \frac{1}{T_{Accelerated}}\right)}$$

By using these MTTFs at 85°C and 95°C, we can easily calculate the activation energy (Q) and from this activation energy we can proceed to obtain an acceleration factor (AF) for these tests, and then calculate the MTTF at the body temperature. Moreover, after analyzing our failed samples we have found out and mentioned in the past progress reports that some of the samples at the 95°C tests have failed prematurely due to the enhanced dissolution rate for silicon at this temperature. Since the dissolution reaction is an exponential function of temperature, the samples at the 85°C tests have been effected less than the ones at 95°C. The model we use only accounts for acceleration of moisture diffusion, but not dissolution. We will still keep and update the data for the tests performed at 85°C. Moreover, for our calculations we assume that

all the samples in the 85°C tests have also failed the same time as the longest going sample in the 95°C tests and proceed with the calculations as follows:

$$MTTF|_{85°C} = 257.6\,Days \quad MTTF|_{95°C} = 118.7\,Days$$

Q=0.88 eV, AF(95°C)=179.5, AF(85°C)=82.7

$$MTTF|_{37°C} = 58.4\,Years$$

We should also note that we have included every single sample in the 85°C and 95°C soak tests in this calculation except the 15% which failed during the first day (we assume that these early failures can be screened for). Moreover, some of these capsules have failed due to mishandling during testing rather than due to actual leakage into the package. If we disregard the samples that we have attributed failure due to mishandling we obtain a longer mean time to failure:

$$MTTF|_{85°C} = 396\,Days \quad MTTF|_{95°C} = 136\,Days$$

Q=1.217 eV, AF(95°C)=1304, AF(85°C)=447

$$MTTF|_{37°C} = 485\,Years$$

2.1.3 New Accelerated Soak Tests in Phosphate Buffered Saline

We have continued our accelerated soak tests with the ultrasonically machined glass capsules bonded to silicon substrates. In these tests, after bonding we have coated the interface between the glass capsule and the polysilicon bond with a biocompatible silicone coating to prevent the dissolution of the polysilicon. This coating is working effectively after 40 days. Tables 4 and 5 below summarize the status of these packages. We have started with 8 devices each at 85°C and 95°C. Out of the original 8 packages at the 85°C tests, one of them failed after one day (premature failure) due to a fault on the bonding surface. The remaining 7 are still dry and being tested. Out of the 8 devices in the 95°C tests, we have one failure after one day (sample is excluded from calculations) due to a scratch on the bonding surface. The other 7 devices at this temperature are still dry and under test. In the previous set of soak tests performed at 95°C (without any coatings), our mean time to failure was 38 days. In the present set, even after 40 days we did not observe any dissolution indicating the effectiveness of the coating.

The packages in these accelerated tests have been monitored every few days for room temperature condensation both electrically by means of an integrated dew point sensor and also visually by the aid of a microscope. We define failure as room temperature condensation. With these ultrasonically machined glass capsules, due to their flat top surface, we have the additional advantage of being able to monitor their bonding surface and the glass capsule to polysilicon interface for discoloration and dissolution. We have found that even though the saline solution is refreshed daily to maintain a constant concentration, at these high temperatures without a

silicone coating it is simply very hard to prevent the dissolution of the polysilicon layer. We are looking into other ways to inhibit polysilicon dissolution at these high temperatures. Nevertheless, we will update the results of our soak tests in the coming quarters.

Table 4: Key data for soak tests in saline at 85° C.

Number of packages in this study	8
Soaking solution	Saline
Failed within 24 hours (not included in MTTF)	1
Packages lost due to mishandling	0
Longest lasting packages in this study	40 days
Packages still under tests with no measurable room temperature condensation inside	7
Average lifetime to date (MTTF)	*40 days*

Table 5: Key data for room temperature soak tests in saline 95° C.

Number of packages in this study	8
Soaking solution	Saline
Failed within 24 hours (not included in MTTF)	1
Packages lost due to mishandling	0
Longest lasting packages in this study	40 days
Packages still under tests with no measurable room temperature condensation inside	7
Average lifetime to date (MTTF)	*40 days*

2.1.4 Ongoing Room Temperature Soak Tests in Saline

We have also continued our soak tests in phosphate buffered saline at room temperature. Table 6 lists some of the pertinent data from these soak tests. These tests were started with 6 packages acting as a control study independent of our accelerated tests. One of these packages failed within one day, most likely due to surface defects or poor bonding due to misalignment. Another one failed after 160 days of soaking. The remaining 4 samples are dry and still under test. These samples, similar to our other samples, are tested visually with the aid of a microscope and electrically with the help of dew point sensors integrated into the package substrate. We have calculated an average lifetime of 829 days for these samples with the longest lasting sample in these tests reaching a total of 1005 days.

2.1.5 In-Vivo Tests

This quarter we have sent 8 devices to our colleagues at Hines VA Hospital in Chicago, Illinois (Dr. Lisa Reidy). These packages were made from the ultrasonically machined glass capsules and the silicon packaging substrate containing stimulating electrodes. The packages are an exact mockup of what will be used in future functional microstimulators. We will report on

biocompatibility results from these packages in the next progress report. We also implanted 3 packages at the Kresge Hearing Research Institute at the University of Michigan. Of these, two are implanted into a guinea pig for 1 month and a third one in another guinea pig for 2 months. These devices were placed on top of the dura, in an area where the skull has been cut away. The skin was then sutured leaving the devices sitting on top of the dura. This placement is compatible with the eventual placement of the hermetically-packaged telemetry platforms for application in the CNS. This quarter 2 of these devices have been harvested. After an implantation duration of 1 month both packages came out intact and did not have any indication of moisture inside them. We have carefully inspected the samples under a microscope and have seen no indication of moisture or any stains caused by exposure to the body fluids on the bonding surface. The measurements from the dew point sensors inside the package also indicate that there is no moisture inside the packages. In both of these implants we have noticed that some bone and skin has grown around them. There was no sign of infection in the animal and no adverse effects that our histology technician could determine.

Table 6: Data for room temperature soak tests in saline.

Number of packages in this study	6
Soaking solution	Saline
Failed within 24 hours (not included in MTTF)	1
Packages lost due to mishandling	1
Longest lasting packages in this study	1005 days
Packages still under tests with no measurable room temperature condensation inside	4
Average lifetime to date (MTTF)	*828.8 days*

Furthermore, we have inspected the silicon substrate for dissolution, but did not observe any indication of dissolution of the silicon substrate or the polysilicon films at body temperature. There was no sign of etching of any of the thin films used in the package. We should also mention that we did not expect any damage to any of the layers used in the package or to the substrate itself at body temperature, but as mentioned in previous reports, we have observed dissolution at our accelerated soak tests in high temperature in saline. In short, these tests reveal that our present package is robust, biocompatible and stays hermetic during the in-vivo tests. We will report additional results as they become available in the coming quarter.

In addition to these test results, we have also obtained additional in-vivo results from the series of tests conducted at Vanderbilt University in collaboration with Dr. D. Zealear. A more detailed description of all of these test results will be presented in our next progress report once all the histology results are available.

2.1.6. Characterization of Package Mechanical Strength

We had performed tensile pull tests to measure the bond strength of the present package in previous quarters. This quarter, we prepared 10 more packages and increased the sample size of our tensile pull tests. The glass capsules are bonded to silicon substrates with the standard anodic bonding process. Next, the packages are attached to wooden blocks at each end. As a final step, one end of the device is held fixed by the wooden block whereas weights are attached to the other end until the devices fracture. From these tests, values for the fracture stress of the

glass-silicon bond are obtained. Figure 6 shows the summary of these results. The maximum fracture force turned out to be 13.5 pounds which corresponds to a fracture stress of 12 MPa. The average fracture force was 6.6 pounds with the standard deviation being 1.95 pounds. These results fall within the range of data stated in the literature which ranged from 2 to 25 MPa [1,2]. After fracture, the samples are dismounted from the wooden blocks and inspected under SEM. Figures 7 and 8 show the overall and the closeup view from one of these samples. As seen in Figure 7 the package breaks from the lowest resistance point in glass which is expected of a bond which is stronger than both silicon and glass. We should also add that the bond is very good around the perimeter of the package. Figure 8 shows a closeup of the right corner of the same package showing that there is glass residue around the bonding region.

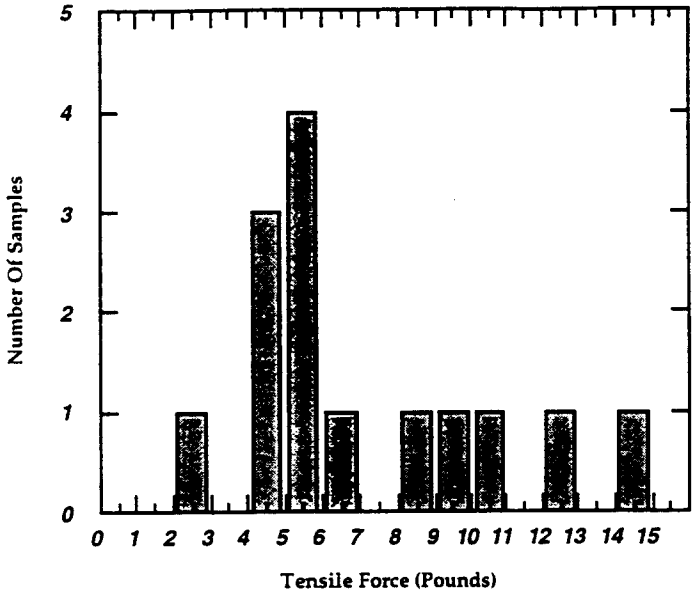

Figure 6: The Summary of Pull Test Results From the Glass Silicon Package.

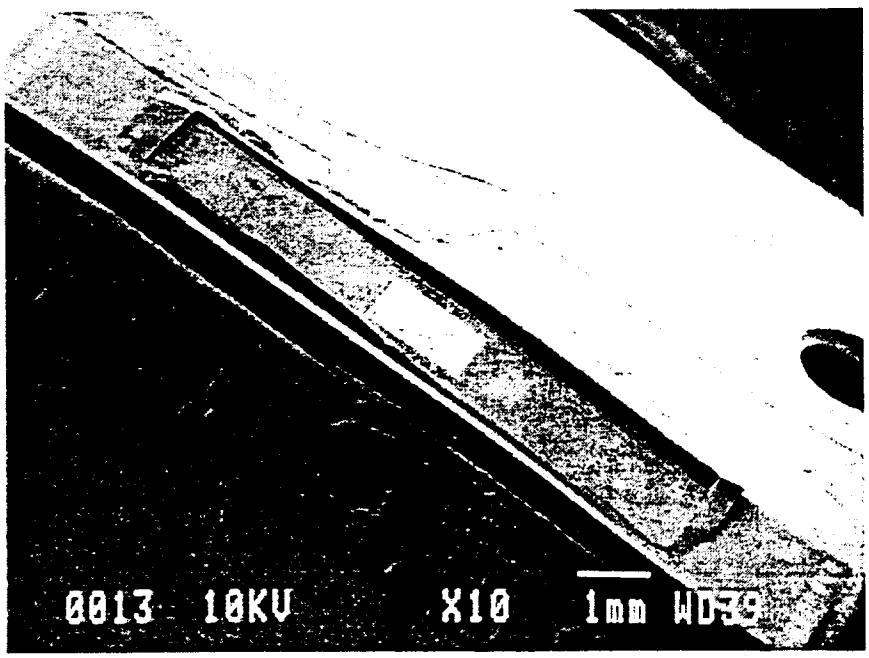

Figure 7: The overall view of a sample package that is pulled apart.

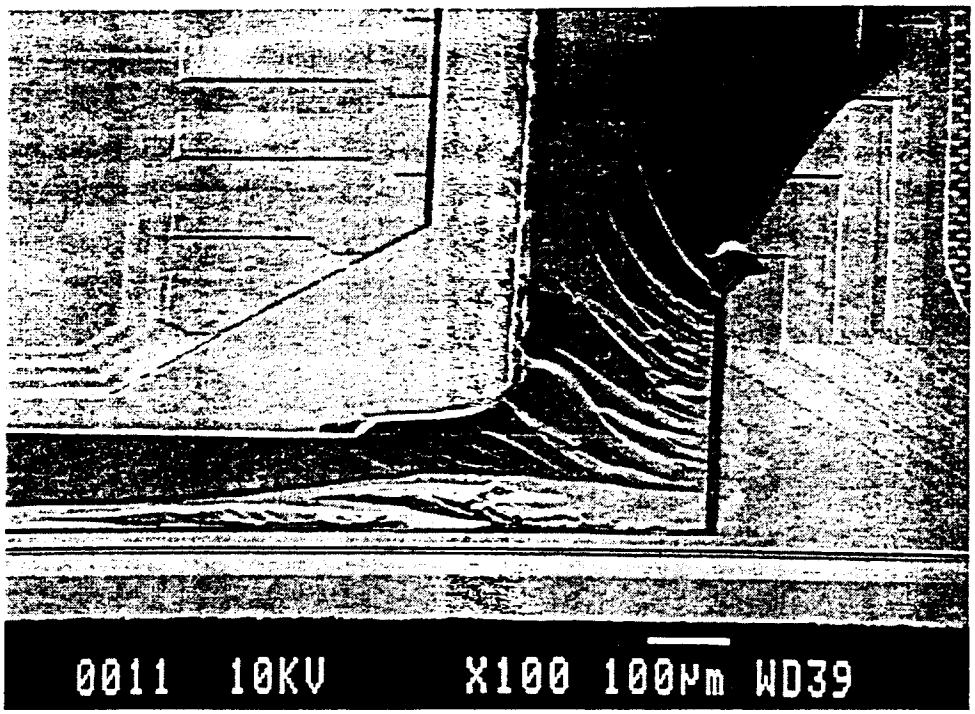

Figure 8: Closeup view of the right hand corner of the package pulled apart.

2.1.7 An All Silicon Package Made From Evaporated Glass Deposited On Silicon

Most of the packages we have made up to now were either made with custom molded glass capsules or ultrasonically machined glass capsules bonded to silicon substrates. In either case, we have used bulk Pyrex glass to create the strong anodic bond. There is current interest in using a thin layer of evaporated glass [3] as an intermediate layer to fabricate an all silicon package. In the past some investigators have utilized sputtered layers of sodium doped glass to generate anodic bonds between glass films and very smooth bulk silicon. However, sputtering glass is a very slow process and is not an economical or feasible approach for many applications. Glass can also be evaporated using a standard process which provides a much higher deposition rate which is more compatible with the 5µm thick layers that are typically required for achieving a good silicon-glass bond. As an alternative to the glass-silicon package, we are also exploring the feasibility of using a silicon-silicon package. This package provides a number of advantages, including the easier machineability of silicon, and the potential for making the package much thinner than is possible with a glass package.

A wafer with a thin layer of evaporated glass has been obtained from a group in Denmark. This wafer was then diced with the same dimensions of a glass capsule and then the individual pieces were cleaned and bonded to a silicon substrate (Figure 9 shows a sample package). Due to the thickness of the glass layer (5 microns), we have utilized a smaller voltage for the electrostatic bond, but the remaining parameters like bonding temperature and time remained the same as in our usual bonding process. Our preliminary experiments indicate that we are generating a bond and have pulled several of the samples that are bonded. Several of the bonds broke without much resistance, but in some others the bonds have resisted a tensile force of 0.8 and 14N/mm^2 which is comparable to results reported [3]. In most of these devices after the bonds come apart we observe that either the glass is pulled out of silicon or the polysilicon

layer come out of the silicon substrate both indicating a strong bond at the polysilicon glass interface. We can immediately observe the quality of the bond from these preliminary experiments. It should be noted, however, that our preliminary experiments also have produced a number of bad bonds, indicating that we do not yet have a reproducible and uniform process to create hermetic bonds. This we believe will be overcome in the future as we better characterize our bonding procedures and are able to obtain evaporated glass films with a better consistency. It should also be mentioned that we are pursuing this approach at a fairly low level of effort and do not intend to replace our current glass-silicon technology with this new approach.

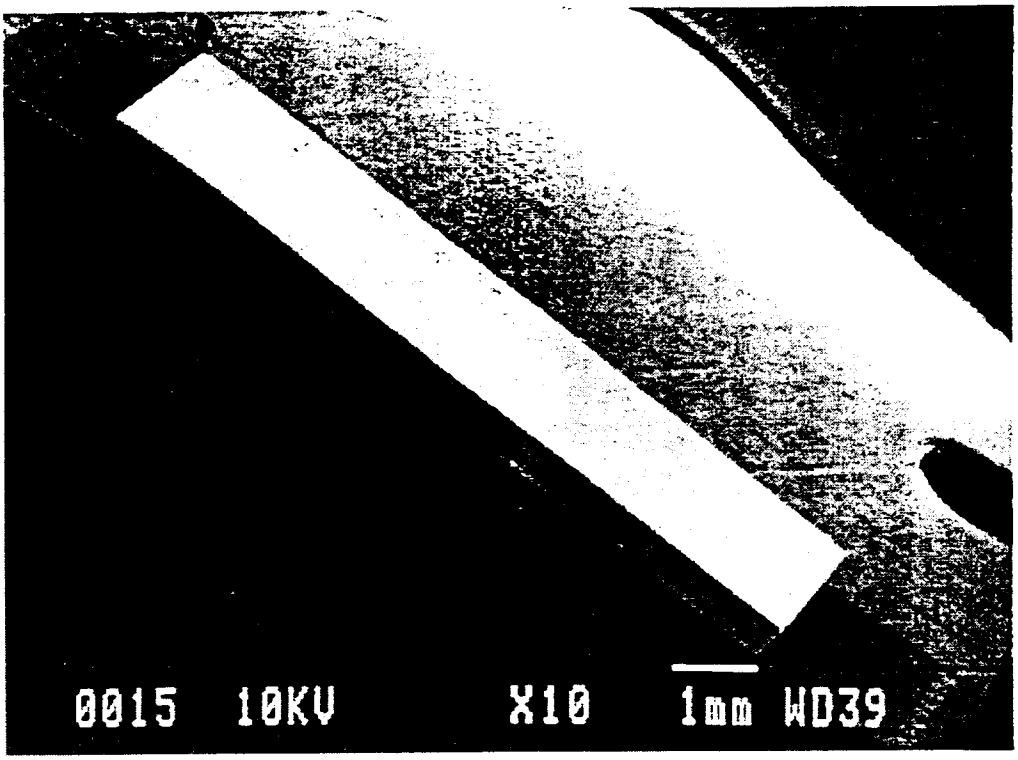

Figure 9: The all Silicon package made with evaporated glass deposited on silicon bonded to Silicon package substrate.

2.2 Packaging and Microtelemetry For Next Generation Microstimulators

The single-channel microstimulator that has been under development for the past few years under funding from the Neural Prosthesis Program, is about 3 orders of magnitude smaller than conventional implantable stimulation units that use hybrid thin-film technology. As part of our contract goals, we are developing miniature packages for a variety of implantable neural prostheses. In order to minimize the size of these packages, we have been examining ways to reduce the volume of implantable stimulators by another order of magnitude. Figure 10 illustrates how this size reduction can be achieved. By far the largest components in the microstimulator system are the charge storage capacitor and the discrete receiver coil. These two components take up about 90% of the total microstimulator volume. As shown in Fig. 10, the volume of the microstimulator can be reduced by an order of magnitude by integrating the receiver coil directly on the CMOS substrate and by not using a charge storage capacitor. While the microstimulator is currently about 2.5 mm thick and has a volume of about 50 mm^3, a system

with an integrated coil and no charge storage capacitor will be in the 300 μm to 500 μm thickness range and have a volume of about 6 mm^3. We call these extremely low volume FES systems mini-microstimulators.

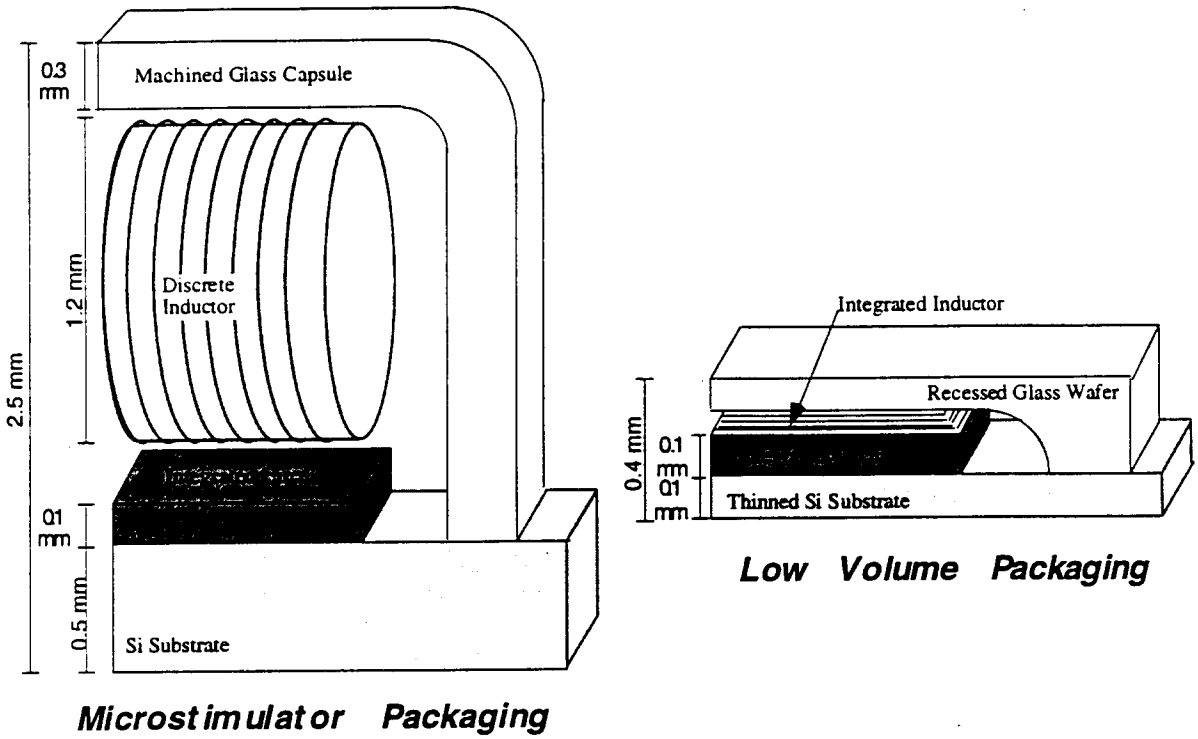

Figure 10: A scale drawing comparing the volume of the microstimulator with the volume of a mini-microstimulator.

Figure 11 shows one of the integrated receiver coils that we have developed for use in mini-microstimulator systems. This coil has 14 turns, dimensions of 2 mm by 10 mm, and has electroplated copper windings and an electroplated nickel-iron (NiFe) core. The coil in Figure 11 has integrated RF receiver circuitry mounted on top of it for telemetry testing. This circuitry includes a 4 MHz tuned RF receiver, a supply voltage generator, a clock recovery circuit, and a data demodulator. The coils and RF receiver circuit have been successfully tested telemetrically, and over 20 mW of power was received at a distance of 3 from the transmitter. Although the circuitry and the coil in Figure 11 are on separate silicon substrates which have been hybrid attached, the coil technology is fully CMOS compatible, and in the future they will be fabricated together.

We are developing a nerve cuff stimulation system to demonstrate the feasibility of a telemetry powered mini-microstimulator. Stimulating nerve cuffs are a well suited application for a mini-microstimulation device because they need relatively low current levels. Typical nerve cuff stimulation levels are 100 μA to 2 mA, which is a good fit with the 3 mA output that is feasible using on-chip coils and no charge storage capacitor (for comparison the microstimulator stimulates with an output of 10 mA or more). Table 7 shows the specifications for the full 8 channel nerve cuff system which we are developing, and Figure 12 shows an exploded view of such a system. Note that the nerve cuff and electrodes can be either integrated directly with the hermetic packaging substrate, or the nerve cuff could be attached by a connector.

This quarter we have completed the physical layout and begun fabrication of the complete eight channel nerve cuff stimulation system. We have also include the latest version of the single channel microstimulator circuitry and the multi-channel microstimulator circuitry on this mask set as well. During the coming quarter we will complete fabrication of all of this circuitry and begin testing the various systems.

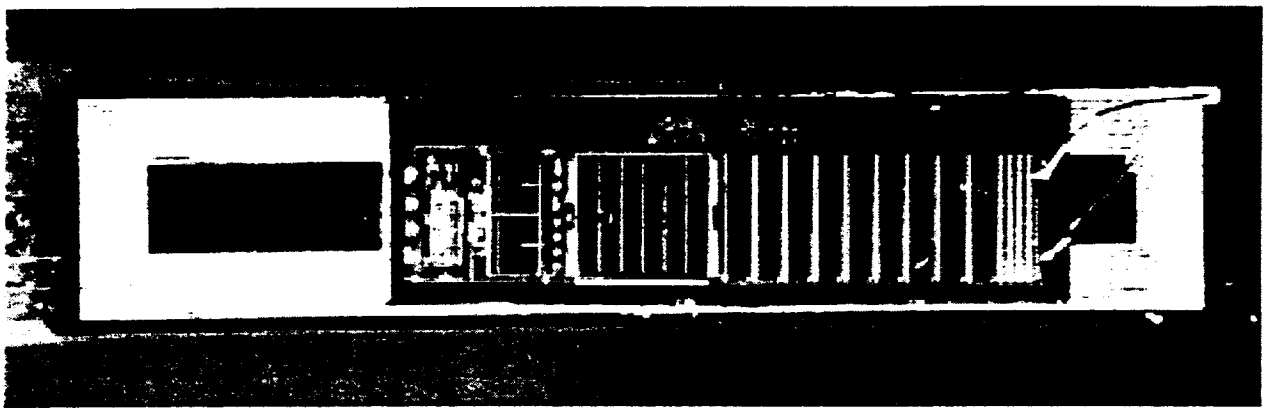

Figure 11: Photograph of the integrated RF receiver circuitry mounted on top of a 2 by 10 mm, 14-turn on-chip. This 1.29 by 5.78 mm circuit includes a 4 volt generator, a data demodulator, and a clock recovery circuit.

Table 7: The specifications for a telemetry powered stimulating nerve cuff.

8-Channel Peripheral Nerve Stimulation System Specifications

General

Dimensions = 2.0 mm X 10 mm X 0.5 mm	Power Delivery = Telemetry
Power Consumption < 12 mW	On Chip Regulated Supply = 4 Volts, Gnd

Telemetry Link

Receiver Coil = On-chip (2.2 mm X 8 mm)	Range = 3 cm
Transmitter Coil = Planar, air core (80 mm dia.)	Carrier Frequency = 4 MHz
Modulation Frequency = 1 kHz to 50 kHz	

Stimulation

Output Channels = 8	Amplitude = 0 to 2 mA (62.5 μA steps)
Duration = 0 to 2047 μS (2 μS steps)	Stimulation Protocol = Bi-phasic
Frequency ≤ 50 Hz	Output Load < 1.9 KΩ

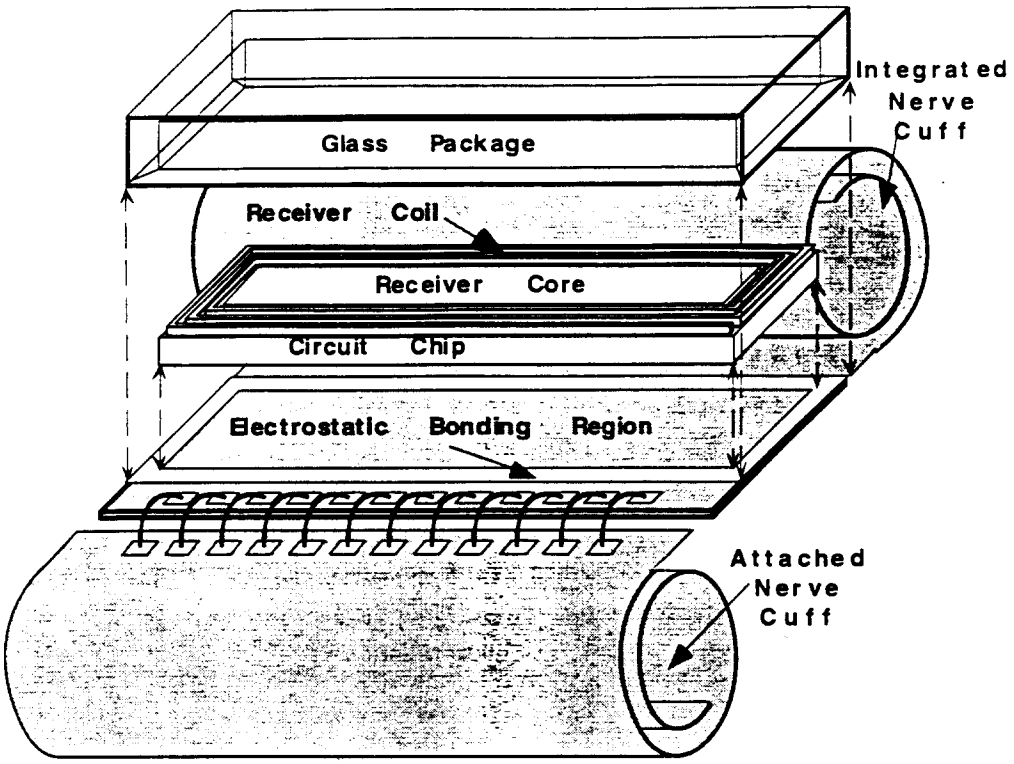

Figure 12: One mini-microstimulator application is this nerve cuff. The nerve cuff can be either integrated directly with the packaging substrate, or attached by a connector.

2.2.1 Layout of the Mini-Microstimulator Circuitry

This quarter we completed the physical layout of the 8 channel nerve cuff stimulation system. The circuitry is 2 mm wide by 8.5 mm long, and it nominally dissipates 4 mW of power (it dissipates a maximum of 12 mW of power when stimulating). The 14 turn, 2 mm by 8 mm RF receiver coil is laid out directly on top of the circuitry, and it will be electroplated there after the circuitry is fabricated. This circuitry contains a total of 2,800 transistors and it is completely integrated, requiring no discrete components at all (such as capacitors or coils). Figure 13 shows a block diagram of the nerve cuff circuitry, and Figure 14 shows the physical layout of the circuitry.

Figure 15 shows the output stimulation waveform, and illustrates the programmable output parameters. The output switches are quite large, so that less than 0.1 volts is dropped across them during maximum stimulation. The programmable current source is also designed so that less than 0.1 volts is dropped across it during maximum stimulation. This means that of the 4 volts generated on chip, 3.8 volts is available to drive the current through the load. This means that the maximum stimulation level of 2 mA can be driven through loads as high as 1.9 kΩ by this system. The regulated current source uses a diode reference to isolate it from the system's 4 V supply. Simulations show that even if the 4 volt supply has a 100 mV ripple, the current source ripple will be kept well below 20 μA. The nerve cuff system requires 40 data bits (and 5 parity bits) to completely program the system output, and these data bits are described in Table 8.

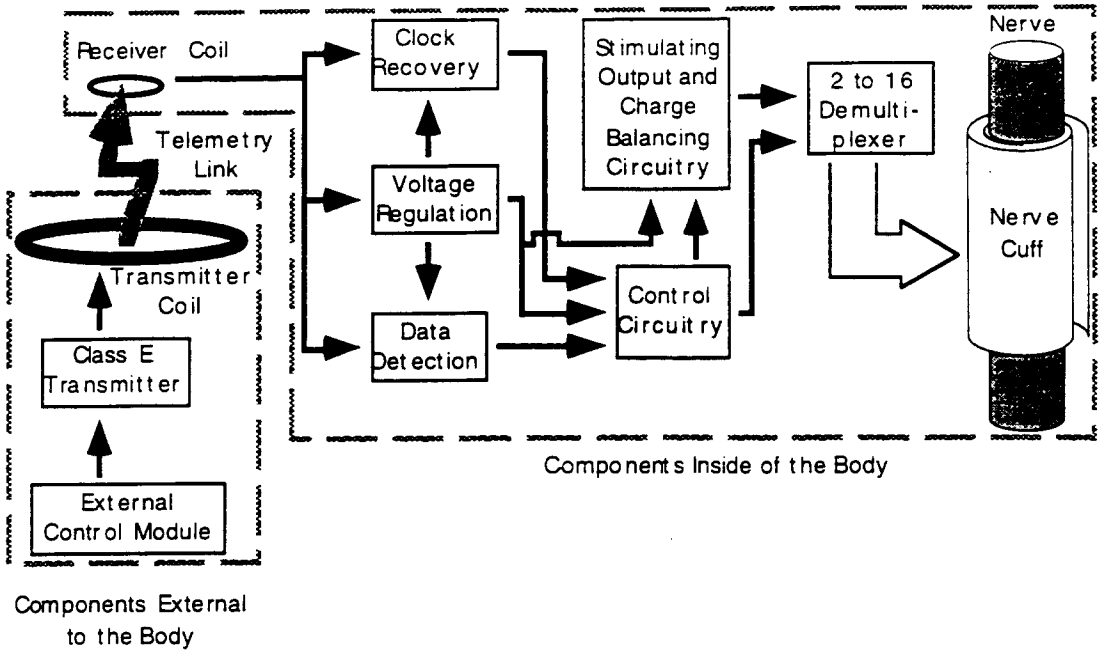

Figure 13: Block diagram of the nerve cuff stimulation system

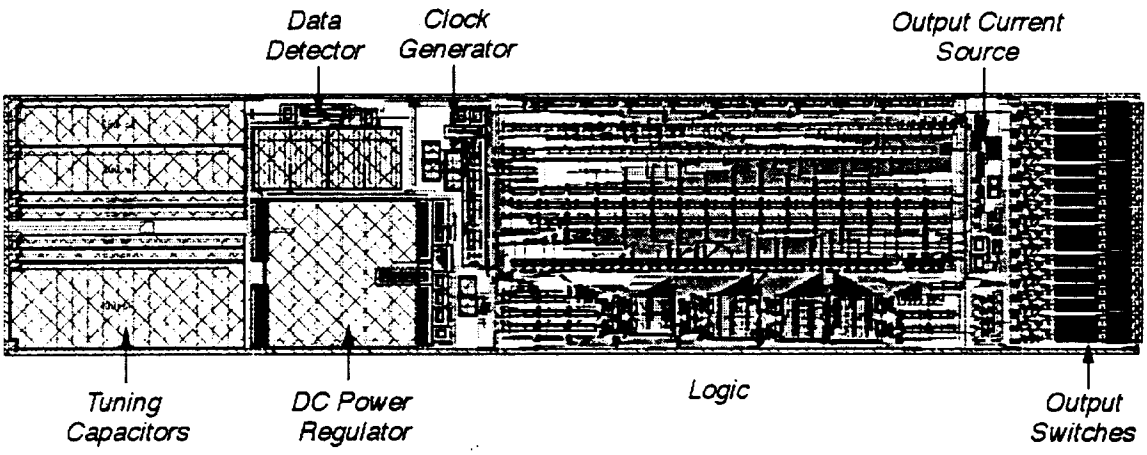

Figure 14: Layout of the nerve cuff stimulation system

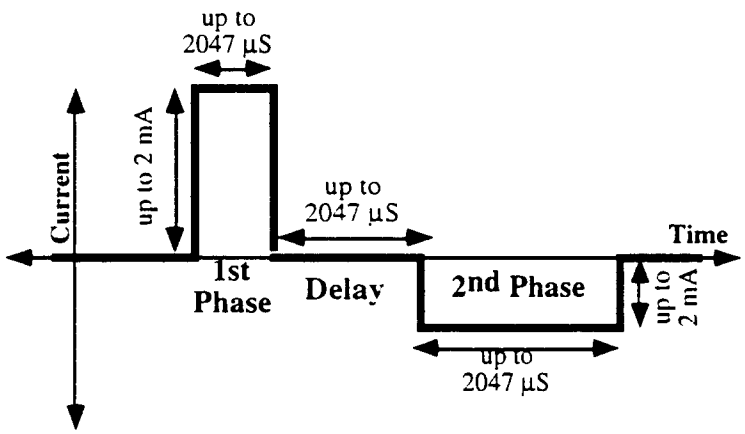

Figure 15: The output stimulation waveform

Table 8: The programmable parameters of the nerve cuff system

Parameter	Range	Required Bits
Address of Device	Selection from up to 8 devices	3 bits
1st Phase Current Magnitude	0 to 2 mA (32 steps of 62.5 μA)	5 bits
2nd Phase Current Magnitude	0 to 2 mA (32 steps of 62.5 μA)	5 bits
Electrode Selection	Selection from 8 electrodes pairs	3 bits
1st Phase Duration	2 to 2048 μS (1024 steps of 2 μS)	10 bits
Inter-phase Delay	2 to 2048 μS (16 steps of 128 μS)	4 bits
2nd Phase Duration	2 to 2048 μS (1024 steps of 2 μS)	10 bits
Parity Bits		5 bits
Total		*45 bits*

3. ACTIVITIES PLANNED FOR THE COMING QUARTER

Our efforts on the various aspects of this project will continue in the coming quarter. First, we will continue soak tests of glass-silicon packages in saline and will acquire additional soak test data to complement that we obtained previously. Testing of the new set of glass capsules that have been coated with silicone will continue to further study the effect of dissolution at these higher temperatures. We will continue to develop techniques for preventing the dissolution of polysilicon at higher temperatures so that long-term accelerated tests could continue to be conducted.

In the area of microstimulator development, in the coming quarter we will complete the fabrication of the nerve cuff circuitry and begin testing. We have also include the latest version of the single channel microstimulator circuitry and the multi-channel microstimulator circuitry on this mask set, and we will test these devices as well as soon as we have completed fabrication.

Finally, we will continue to work with a number of groups that have been interested both in microstimulators and in our packaging technology, including Vanderbilt University, VA Hines Hospital, Case Western Reserve University, Johns Hopkins University. Results from these collaborations will be reported as they become available.

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[3] R. Reus and M. Lindahl, "Si-to-Si wafer bonding using evaporated glass", *Digest, Int. Conf. on Solid-State Sensors and Actuators, Transducers '97*, pp. 661-64, June 1997